

"Express Mail" mailing label number: EV 320 857 705 US

Date of Deposit: December 4, 2003

Our Case No. 09800240-0068

IN THE UNITED STATES PATENT AND TRADEMARK OFFICE
APPLICATION FOR UNITED STATES LETTERS PATENT

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TITLE: MICROPARTICLES

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MICROPARTICLES

BACKGROUND

Rapid advances in biotechnology have led to the discovery of numerous protein and peptide therapeutics, many of which have recently reached the marketplace or are currently under regulatory review by the United States Food and Drug Administration. Unlike traditional small-molecule drugs, however, proteins and peptides generally cannot be administered orally; injection or infusion is most often required. Further, because of their fragility and short *in vivo* half-lives, encapsulation of proteins in biodegradable polymeric devices, from which the drug can be delivered, locally or systemically, for a prolonged period of time, has been a promising and intensely studied solution to these problems.

Biodegradable microparticles containing a variety of polymers have been the most studied delivery vehicle due to relatively simple fabrication and facile administration to a variety of locations *in vivo* through a syringe needle. One particularly preferred microparticle material is chitosan, a hydrophilic biodegradable natural polymer that has been used for drug delivery systems in recent years. To manufacture microparticles of chitosan for this type of application, various methods have been employed, including spray drying and classic emulsions, both at the bench and industrial scales. However, neither technique has yielded uniform microparticles and microcapsules (to be collectively referred to as microparticles) with precisely controlled size and size distribution. In fact, standard deviations equal to 25-50% of the mean diameter are not uncommon.

Control of particle size and size distribution has several important implications for controlled-release drug delivery. For example, there typically is an ideal particle size that provides a desired release rate and route of administration. Microparticles that are "too small" exhibit poor encapsulation efficiency, may migrate from the site of injection, and may exhibit undesirably rapid release of the drug. Particles that are "too large" may not easily pass through

a syringe needle. Thus, the typical polydisperse particles generated by conventional fabrication techniques must be filtered or sieved to isolate particles within the desired size range, and the particles outside that range are wasted.

Moreover, with traditional technologies for spraying microdroplets from nozzle-type devices, the minimum particle size typically obtainable is limited by the size of the nozzle opening. Usually, it is not possible to make drops smaller than the nozzle opening; typically, droplet diameters are 1-4 times the diameter of the nozzle. This presents several difficulties as the desired particle size decreases. One problem is that fabrication of the nozzles themselves becomes more difficult as size decreases.

A second limitation stems from the pressure needed to pump fluids through small nozzles. The pressure required scales with R^{-4} , wherein R is the radius of the nozzle. Thus, pumping virtually any liquid through a nozzle of 5- μm diameter would require special equipment, if it could be done at all. Also, some compounds to be encapsulated, such as plasmid DNA, may be damaged by shear forces. In general, the damage is approximately inversely proportional to the diameter of the orifice. Thus, decreasing the nozzle diameter from 100 to 5 μm would increase the damage done to any encapsulated compound by a factor of 20.

Published U.S. Patent Application No. 2002/0,054,912 (published May 9, 2002, entitled "Microcapsules", to Kim et al.) hereby incorporated by reference, teaches a process wherein micro-and nano-sized particles, preferably spherical, are produced by pumping material, usually a polymer dissolved in an organic solvent, through a small orifice and then shaking the liquid with an acoustic type-wave, where the velocity of the fluid is increased beyond the velocity produced by pressure behind the liquid. In this process, the nozzle diameter may be larger than the particles produced. For example, 5- μm droplets can be prepared from a much larger nozzle, such as a nozzle of 100 μm diameter. The droplets are collected in a solution, and the presence of a surfactant prevents the droplets from sticking together before the evaporation of the organic solvent leads to the hardening of the droplets into microparticles.

The pressures needed to form very small particles are reduced to ranges easily obtained with commercial high-pressure pumps such as those commonly supplied with high-pressure liquid chromatography systems. Furthermore, the shear forces are greatly reduced for a given particle size, and the difficulties encountered with very small diameter nozzles are also eliminated. Aspects of the invention are described in "Fabrication of PLG microspheres with precisely controlled and monodisperse size distributions" J. Controlled Release 73(1):59-74 (May 18, 2001).

The vibration or shaking can be achieved by, for example, a piezoelectric transducer driven by a wave generator, and breaks the stream into a train of uniform droplets. Droplet size is determined by the orifice diameter, the solution flow rate, the vibration frequency and amplitude. Thus, by varying these four parameters droplet size can be controlled.

The velocity of the fluid is increased beyond the velocity produced by the pressure behind the liquid by employing an additional downward force that will 'pull' the liquid jet through the orifice, reducing the jet size below the diameter of the orifice. One example is an electrohydrodynamic technique in which electrical forces act to reduce the diameter of the liquid jet and the resulting droplets. The electrohydrodynamic technique is activated through injection of charge of desired polarity into the liquid by applying a high voltage either to the nozzle or directly into the liquid, for example, with a battery, or with a transformer and a rectifier to convert standard current. Outwardly directed electrical tension forces result at the charged liquid meniscus of the nozzle opening, enabling a smaller drop to fall from the nozzle (the "drip mode"). The reason for this reduction in drop size is believed to be that there are two forces present, gravitational and electrical, that are working together to pull the liquid off of the nozzle, while surface tension forces hold the liquid at the nozzle. As the amount of charge injected increases, the electrical tension forces accordingly increase, eventually dominating the gravitational and surface-tension forces and reducing the drop size. Further increase in charge injection beyond a certain threshold value results in very powerful electrical tension forces that literally pull the liquid out of the nozzle to

form a thin charged liquid jet, which in turn breaks up into fairly uniform droplets (known as the “jet mode”). Jet mode changes from single-jet to multi-jet mode as charge injection is further increased.

5 Another example of an additional downward force employed is a separate liquid stream (typically immiscible) through the orifice, adjacent and parallel to the particle-forming liquid, at a velocity greater than the particle-forming liquid. The particle-forming liquid is pulled along by the drag forces at the liquid/liquid interface. The particle-forming jet is reduced in diameter by a factor that is proportional to the difference in linear velocities of the two streams.

BRIEF SUMMARY

In a first aspect, the present invention is a method of forming particles, comprising accelerating a first stream comprising a first liquid, applying a
5 charging voltage of at most 1.5 kV to the first stream, and vibrating the first stream, to form particles.

In a second aspect, the present invention is a method of forming chitosan particles, comprising accelerating a first stream comprising an aqueous solution of chitosan, applying a charging voltage of at most 1.5 kV to the first stream,
10 vibrating the first stream, to form particles; and maintaining the particles at a pressure of at least 0.1 mm Hg and of at most 760 mm Hg, while heating the particles to a temperature within ± 50 °C of the boiling point of water at the pressure. The accelerating comprises contacting the first stream with a second stream, and the second stream comprises a hydrophobic liquid.

15 In a third aspect, the present invention is particles comprising chitosan having an average diameter of at least 50 μm to at most 100 μm . 90% of the particles have a diameter that is within 2% of an average diameter of the particles.

In a fourth aspect, the present invention is particles comprising chitosan having an average diameter of at least 1 μm to 50 μm . 90% of the particles have a
20 diameter that is within 1 μm of an average diameter of the particles.

In a fifth aspect, the present invention is a method of forming gelatin particles, comprising accelerating a first stream comprising an aqueous solution of gelatin, applying a charging voltage of at most 1.5 kV to the first stream, and
vibrating the first stream, to form particles.

25 BRIEF DESCRIPTION OF THE DRAWINGS

Figure 1 is an illustration of an apparatus for fabricating the microparticles.

Figure 2 is a graph of the size distribution of a batch of 60 μm chitosan microparticles.

30 Figures 3 (A) and (B) are SEM micrographs of 60 μm chitosan microparticles.

Figures 4 (A) and (B) are SEM micrographs of cross-sections of 60 μm chitosan microparticles.

5 Figure 5 is SEM micrographs of gelatin microspheres of isoelectric pH = 5.0 prior to crosslinking. The scale bar = 20 μm . The diameter of the microspheres are (A) 15 μm . (B) 30 μm . (C) 45 μm .

Figure 6 is SEM micrographs of gelatin microspheres of isoelectric pH = 9.0 prior to crosslinking. The scale bar = 20 μm . The diameter of the microspheres are (A) 15 μm . (B) 20 μm . (C) 45 μm .

10 Figure 7 is light micrographs of gelatin microspheres after crosslinking. Initial diameter of microspheres after swelling in aqueous media: (A) 50 μm (10x). (B) 50 μm (40x). (C) 100 μm (10x). (D) 150 μm (10x).

Figure 8 illustrates a blade chopping 150 μm droplets into smaller droplets of: (A) 142- μm -diameter and 80- μm -diameter. (B) 149- μm -diameter and 40- μm -diameter.

DETAILED DESCRIPTION

The technique described above is not best suited to the fabrication of chitosan particles. Chitosan dissolves in acidic water solution, and is thus delivered in aqueous droplets that are collected in a bath of an immiscible solvent. However, the presence of a surfactant does not separate the droplets as reverse micelles, and the chitosan droplets cluster before hardening into microparticles. The present invention provides a solution to this problem based on the discovery that charging the droplets prevents them from coalescing before hardening, thus providing a microparticle fabrication method especially suitable for hydrophilic polymers such as chitosan.

A chitosan solution is prepared by dissolving chitosan in an acidic aqueous solvent, and the solution is loaded in the apparatus **100** illustrated in Figure 1. A water-immiscible solvent, such as a vegetable oil, is also loaded in the apparatus, and a collection bath **102** is loaded with an immiscible solvent. A stream of the chitosan solution is generated by an inner nozzle **104** and charged with a charging voltage, while a carrier stream of the immiscible solvent is generated by an outer nozzle **106** coaxial with the inner nozzle. The diameter of the inner nozzle is preferably greater than $\frac{1}{2}$ of the average diameter of the particles, and more preferably at least equal to the average diameter of the particles. The carrier stream surrounds the inner stream of chitosan solution and flows through the outer nozzle with it. As described above, the carrier stream imposes frictional force on the inner chitosan stream to pull and accelerate it, and a liquid jet of chitosan solution thinner than the size of the inner nozzle orifice can be generated by adjusting the flow rate of the carrier stream. Acoustic excitation of a desired frequency is then applied to the stream, disrupting it to form a train of uniform droplets.

Further reduction in the droplet size can be achieved by placing a blade or a nanowire in front of a droplet exiting the nozzle so that the blade or the nanowire is placed in the flowing stream, thus chopping the droplet into two smaller droplets of equal or different sizes. When the two smaller droplets are of different size, the smaller of the two droplets can be made with a radius smaller than 5 μm ,

depending on the sharpness of the blade, the position of the blade, or the size of the nanowire

The droplets are collected in the collection bath of an immiscible solvent, where the temperature of the bath is such as to induce evaporation of the aqueous solvent. The droplets do not aggregate due to the coulombic repulsion between their electric charges, and harden into chitosan microparticles. The microparticles are collected from the solvent, washed, dried and optionally treated with alkali to eliminate residual acidity from the chitosan solution.

The average diameter of the microparticles is at most 100 μm . More preferably, the average diameter of the microparticles is at most 50 μm . Yet more preferably, the average diameter of the microparticles is at least 1 μm and at most 50 μm . Most preferably, the average diameter of the particles is at least 10 nm and at most 50 μm . If the average diameter of the microparticles is at least 50 μm and at most 100 μm , 90% of the microparticles have a diameter that is within 2% of the average diameter of the microparticles. If the average diameter of the microparticles is at least 1 μm and at most 50 μm , 90% of the microparticles have a diameter that is within 1 μm of the average diameter of the microparticles.

The chitosan solution may be prepared by dissolving chitosan in an aqueous solvent containing one or more acids such as formic acid, acetic acid, propionic acid, trifluoroacetic acid, trichloroacetic acid, nitric acid, sulphuric acid, phosphoric acid, hydrofluoric acid, hydrochloric acid, hydrobromic acid, hydroiodic acid, perchloric acid and perbromic acid.

The applied charging voltage is preferably at least 0.5 kV and at most 1.5 kV, more preferably of at least 0.75 kV and at most 1.25 kV, and most preferably of at least 0.9 kV to at most 1.1 kV. Voltages such as those used in electrohydrodynamic droplet generating methods, generally in the 10 kV to 20 kV range, are likely to damage the pharmaceutical compositions contained in the droplets.

The solvent of the collection bath may contain one or more fluids immiscible with water, for instance synthetic or natural oils such as canola oil, corn oil, peanut oil, linseed oil, sunflower oil, soybean oil, flax oil, olive oil,

grape-seed oil and coconut oil. The bath may also contain additional synthetic or natural excipients. The temperature of the bath is preferably from at least 90 °C to at most 170 °C, more preferably from at least 110 °C to at most 150 °C, and most preferably from at least 125 °C to at most 135 °C. The evaporation may be
5 conducted at lower temperatures by keeping the droplets at sub-atmospheric pressures, thus lowering the boiling point of the aqueous solvent. For instance, the evaporation may be conducted at a pressure of 0.54 atm, where the boiling point of water is at 84.4 °C.

Where heat-driven evaporation of the aqueous solvent is not preferred, for
10 instance in the preparation of gelatin microspheres, the collection bath is kept at temperatures in the 0 to 25 °C, in order to promote spontaneous gelation of the aqueous solution. Optionally, a surfactant may be added to the aqueous solution or the collection bath.

After the hardening, the resulting microparticles are filtered, washed with a
15 solvent such as ethanol to remove the oil, and optionally treated with a base to deprotonate their surface. A final washing in water may follow, after which the particles may be lyophilized. Also, the microparticles may be hardened by treatment with a crosslinking agent.

Chitosan microcapsules of controlled size and shell thickness may also be
20 fabricated by inserting a coaxial nozzle inside the nozzle carrying the chitosan solution described above, yielding a stable, hollow jet of chitosan solution surrounding a coaxial jet of a gas or an immiscible liquid. A third, external carrier stream of an immiscible solvent as described above may be applied to the jet, obtaining multilayered droplets that are then hardened in the collection bath,
25 yielding chitosan microcapsules containing the gas or the immiscible liquid.

The technique may also be applied for the fabrication of particles of hydrophilic polymers other than chitosan and gelatin, such as carboxy methyl cellulose, dextran, hydroxypropyl cellulose, poly(acrylamide), poly(acrylic acid), poly(allylamine) hydrochloride, poly(diallyldimethylammonium chloride),
30 poly(N,N-dimethyl acrylamide), poly(ethylene glycol), poly(ethylene oxide), poly(maltotriose), poly(methacrylic acid), poly(N-isopropyl acrylamide),

poly(propylene glycol), poly(styrene carboxylic acid), poly(styrene sulphonic acid), poly(styrene sulphonate), poly(vinyl acetate), poly(vinyl alcohol), poly(vinyl butyral), poly(2-vinyl-N-methyl pyridinium iodide), poly(4-vinyl-N-methyl pyridinium), poly(2-vinyl pyridine), poly(2-vinyl pyridinium bromide),
5 poly(vinyl pyrrolidone), p(methyl vinyl ether), hydroxyalkyl starches such as hetastarch, alkylcellulose, hydroxyalkylcellulose, hydroxyarylcellulose, alginate, and copolymers thereof. Additionally, more than one polymer or copolymer may be used in order to obtain microparticles made of composite materials.

Additionally, the gelated microparticles may be treated with a cross-linking
10 agent in order to modify, for instance, their mechanical properties and solubility. Examples of cross-linking agents include: aldehydes, such as glyceraldehyde, glutaraldehyde, dextran dialdehyde, and carbohydrates; diols, such as ethylene glycol, di(ethylene glycol), polyethylene glycol, propylene glycol, di(propylene glycol), polypropylene glycol; unsaturated diesters such as ethylene glycol
15 dimethacrylate, di(ethylene glycol) dimethacrylate, poly(ethylene glycol) dimethacrylate, poly(lauryl methacrylate-co-ethylene glycol dimethacrylate), propylene glycol dimethacrylate, di(propylene glycol) dimethacrylate, poly(propylene glycol) dimethacrylate; dihydrazides such as malonic dihydrazide, ethylmalonic dihydrazide, succinic dihydrazide, glutaric dihydrazide, adipic
20 dihydrazide, isophthalic dihydrazide, oxalyl dihydrazide, pimelic dihydrazide, 3,3'-sulfonyldibenzene sulfonic dihydrazide; diisocyanates such as m-xylylene isocyanate, 4-methyl-m-phenylene diisocyanate, 2-methyl-m-phenylene diisocyanate, 3,3'-dimethoxy-4,4'-biphenylene diisocyanate, 4-Br-6-methyl-1,3-phenylene diisocyanate, 4-Cl-6-methyl-1,3-phenylene diisocyanate, toluene 2,4-
25 diisocyanate, 1,3-phenylene diisocyanate, 1,4-phenylene diisocyanate, 2,4,6-trimethyl-1,3-phenylene diisocyanate, 1,4-diisocyanatebutane, 1,6-diisocyanatehexane, 1,8-diisocyanateoctane, isophorone diisocyanate; carbodiimides such as N,N-(3-dimethylaminopropyl)-N-ethyl carbodiimide (EDC); salts, such as CaCl_2 ; divinylsulfone, sulfonylurea, hydrolysable
30 polyrotaxane, L-lysine methyl ester, and genipin. Cross-linking may also be carried out or aided by one or more enzymes.

The microparticles of the invention may include proteins and peptides such as human growth hormone, bovine growth hormone, erythropoietin, thrombopoietin, tissue plasminogen activator and derivatives, insulin, monoclonal antibodies (e.g., anti-human epidermal growth factor receptor2 (HERCEPTINTM), anti-CD20 (RITUXIMABTM), anti-CD 18, anti-vascular endothelial growth factor, anti-IgE, anti-CD 11a) and their derivatives, single-chain antibody fragments, human deoxyribonuclease I (DORNASE ALFATM, PULMOZYMETM), type- 1 interferon, granulocyte colony-stimulating factor, leutenizing hormone releasing hormone inhibitor peptides, leuprolide acetate, endostatin, angiostatin, porcine factor VIII clotting factor, interferon alfacon-1, pancrelipase (pancreatic enzymes) and the like.

Hormones and steroids (corticosteroids) useful for the practice of this invention include but are not limited to norethindrone acetate, ethinyl estradiol, progesterone, estrogen, testosterone, prednisone and the like.

Chemotherapeutics useful for the practice of this invention include but are not limited to taxol (PACLITAXELTM), vinblastine, cisplatin, carboplatin, tamoxifen and the like.

Non-steroidal anti-inflammatory drugs (NSAIDs) useful for the practice of this invention include but are not limited to piroxicam and the like.

Vaccine components useful for the practice of this invention include but are not limited to Hepatitis B, polio, measles, mumps, rubella, HIV, hepatitis A (e.g., HAVRIXTM) and the like.

Analgesics useful for the practice of this invention include but are not limited to aspirin, acetaminophen, ibuprofen, naproxen sodium and the like.

Antibiotics useful for the practice of this invention include but are not limited to amoxicillin, penicillin, sulfa drugs, erythromycin, streptomycin, tetracycline, chlorithromycin, ciprofloxacin, terconazole, azithromycin and the like.

Anti-depressants useful for the practice of this invention include but are not limited to ZOLOFTTM, fluoxetine (PROZACTM), paroxetine (PAXILTM),

citalopram, venlafaxine, fluvoxamine maleate, imipramine hydrochloride, lithium, nefazodone and the like.

Other biopharmaceutical compounds useful for the practice of the instant invention include but are not limited to sildenafil (VIAGRATM), acyclovir, gancyclovir, fexofenidine, celecoxib (CELEBREXTM), rofecoxib (VIOXXTM), androstenedione, chloroquine, diphenhydramine HCl, buspirone, doxazocin mesylate, loratadine, clorniphine, zinc gluconate, zinc acetate, hydrocortisone, warfarin, indinavir sulfate, lidocaine, novacaine, estradiol, norethindrone acetate, Medroxyprogesterone, dexfenfluramine, Dextroamphetamine, Doxycycline, thalidomide, fluticasone, fludarabine phosphate, etanercept, metformin hydrochloride, hyaluronate, tetrazocin hydrochloride, loperamide, ibogaine, clonazepam, ketamine, lamivudine (3TCTM), isotretinoin, nicotine, mefloquine, levofloxacin, atorvastatin (LIPITORTM), miconazole nitrate (MONISTATTM), ritonavir, famotidine, simvastatin (ZOCORTM), sibutramine HCl monohydrate, ofloxacin, lansoprazole, raloxifene (EVISTATM), zanamivir (RELENZATM), oseltamivir phosphate, 4-phenylbutyric acid sodium salt, chlorpromazine, nevirapine, zidovudine, cetirizine hydrochloride (ZYRTECTM), bisphosphonates such as pamidronate and zoledronate, nifedipine, felodipine, and the like.

Materials which can further be utilized for the practice of this invention can be found in *Physician's Desk Reference 2000*, 54th Edition, ISBN: 1563633302, *AHFS 99 Drug Information*, Amer. Soc. of Health System, ISBN: 1879907917 and U.S. Patent No. 5,019,400 all incorporated herein by reference in their entirety.

The microparticles of the present invention may be used to incorporate other materials, such as salts, metals, sugars, surface active agents, acids, bases, stabilizers and release enhancing agents. In general, these and other materials (such as polymers, biopharmaceutical compounds and other ingredients) may be incorporated into the outer shell, into the core, in each shell individually or at various levels of the concentric shells. Heterogenous mixture of chitosan and pharmaceutical compounds may also be included.

The microparticles of the invention may be used for many biomedical applications, such as the following:

Passive targeting of phagocytosis. Cells of the immune system, especially macrophages and dendrocytes, are targets for immunization. These “professional” antigen-presenting cells (APCs) can elicit a desired T-cell response to vaccine components. APCs are typically capable of phagocytosis of particles in the range of 1 to 10 μm ; most other types of cells in the body cannot internalize such large particles. By generating in this size range particles containing vaccine components, one can passively target delivery of the vaccine to APCs. Current technologies allow formation of particles of this size, but only with significantly broad size distributions. This methodology allows generation of vaccine-encapsulating chitosan microparticles in which essentially 100% of the particles are of the desired size. APC targeting can be optimized with no waste of expensive vaccine components in microparticles that are too large or too small.

Capillary embolization. Blood vessel size varies throughout the body with the smallest vessels being the capillaries. Tumor tissues typically exhibit particularly tortuous capillary beds that feed the rapidly growing cancer cells. One approach to targeting tumor tissues is to deliver a therapeutic encapsulated in microparticles of a size such that they can become lodged in narrow capillaries. The microparticles are then injected into an artery feeding the tumor. Particles which are too large will become lodged in larger vessels and the therapeutic may be blocked from reaching all of the tumor. Particles which are too small will pass through the tumor and could deliver the therapeutic to other healthy tissue; a serious problem since many anti-cancer agents are, of course, cytotoxic. The availability of microparticles of a precise size will allow the maximization of the number of particles lodged in the smallest capillaries and, therefore, the amount of therapeutic delivered to the desired location.

Targeting by nano-particles. Biodegradable polymer particles less than 1 μm in diameter (nano-particles or nano-capsules) have several specific applications in drug delivery. For example, particles less than 150 nm in diameter

can be targeted for receptor-mediated endocytosis by many different types of cells. Nanoparticles are also useful for oral administration wherein the particles can traverse the intestinal epithelium and be taken up by the M-cells and Peyer's patches. Particles that are too large are poorly taken up, particles that are too small contain little therapeutic. Most current methods for biodegradable polymer nanoparticle fabrication, of which there are many, result in wide variations in particle size. For example, phase inversion nanoencapsulation, in which polymer and drug are dissolved in a solvent (e.g., methylene chloride) and subsequently poured into a non-solvent phase (e.g., petroleum ether), yields particles ranging from 100 to 5000 nm. In contrast, the controlled size distribution will allow optimal nanoparticle targeting.

Control of drug release rate from solid microparticles. Many controlled-release drug delivery systems include microparticles formed from biodegradable polymers. Therapeutic compounds encapsulated in the microparticles are released as the polymer erodes, most often by hydrolysis, in the aqueous environment within the body. The drug release rate depends on several factors such as polymer composition, polymer molecular weight, microparticle porosity, etc. An especially critical parameter is particle size. In general, smaller particles with larger surface and surface-to-volume ratio erode and release therapeutic compounds more quickly than larger particles.

The ability to precisely control microparticle size distribution allows for unprecedented control over drug release rates. By varying liquid flow rate, vibration frequency and amplitude, and charging potential, particle size can be controlled. Furthermore, by continuously and smoothly varying such parameters, pre-defined size distributions can be generated. Since drug release rate depends on particle size distribution, an appropriate size distribution that will provide a desired release rate profile (e.g., zero-order release, pulsatile release, etc.) can then be determined.

Fabrication of composite core/shell microparticles. With the multi-nozzle methodology, composite particles of two or more different materials can be fabricated. Fine control of shell thickness and particle diameter are possible. The

two phases may comprise: a gas core within a chitosan shell; a drug-containing aqueous core within a chitosan shell (the shell could also encapsulate a second or the same compound); a polymer core encapsulating a drug compound within a polymer shell (wherein either the core or the shell may be of chitosan) encapsulating a second or the same compound. Such particles could produce novel and useful drug-release rate profiles.

Fabrication of nanoshells. Virtually any compound can be encapsulated in the interior phase of the particle. If DNA (or other nucleic acids) are encapsulated, “artificial virus” particles that may be particularly useful for gene delivery can be produced.

Pharmaceutical compounds which can be encapsulated/released by the instant invention include but are not limited to nucleic acids, proteins and peptides, hormones and steroids, chemotherapeutics, NSAIDs, vaccine components, analgesics, antibiotics, anti-depressants, and the like.

Nucleic acids useful for the practice of this invention include but are not limited to DNA, RNA, peptide-nucleic acids, oligonucleotides, modified to improve stability (e.g., phosphorothioates, aminophosphonates or methylphosphonates).

EXAMPLES

(1) Uniform chitosan particle fabrication

A 1 % (w/v) chitosan solution was prepared by adding 1 gr of chitosan to 100 mL of 0.2 M aqueous acetic acid and stirring for 3 ~ 4 hours. The solution was filtered through a glass filter and supplied to the main stream tank of the system.

Canola oil was used for the carrier stream and the collection bath. The oil was heated to ~130 °C and supplied to the carrier stream tank and the collection bath. The carrier stream tank and the collection bath were placed on stirrer hot plates to maintain the oil at a temperature > 100 °C while the chitosan solution drops were being generated.

A single nozzle was used to fabricate the solid particles. A high-pressure syringe pump was used to generate the chitosan solution main stream with a flow

rate resolution of 1 $\mu\text{L/hr}$. The carrier stream was pressurized in a container at 30 psi and its flow rate was regulated with a micro-valve in the flow line.

A stable jet with a desired main stream diameter was obtained by applying the appropriate main stream and carrier stream flow rate, and acoustic excitation was applied to the jet. The jet was disrupted into droplets of a size that could be controlled by changing the frequency of the acoustic excitation or the flow rates of the main and carrier stream, respectively.

While the droplets were being generated, a charging voltage of 1 kV was applied, yielding droplets of the same polarity that spread in a canola oil collection bath. The temperature of the collection bath was maintained above 100 $^{\circ}\text{C}$, and the drops hardened through solvent evaporation while staying separated due to coulombic repulsion. The resulting microparticles were left in the collection bath for a total of 1~2 hours. The collection bath containing the chitosan particles was then stirred on a stirrer hot plate for 3 ~ 4 hours to bring water evaporation to completion.

The microparticles were filtered from the oil bath and rinsed with ethanol. Washing in 1 M aqueous sodium hydroxide solution followed in order to deprotonate the surface of the microparticles so that they would not dissolve in water. The particles were then washed with water several times, suspended in water, centrifuged and lyophilized for 2 days.

Figure 2 illustrates the size distribution as determined by a Coulter Multisizer. The SEM micrographs of Figure 3 show the uniform size distribution of 60 μm chitosan particles obtained with the method of the invention, whereas the SEM micrographs of Figure 4 illustrate their high density and homogeneous morphology.

(2) Uniform alginate particle fabrication

The above method used for fabricating chitosan microparticles was also used for fabricating alginate microparticles, except sodium alginate dissolved in water was used instead of the chitosan solution.

The microparticles were cross-linked in 0.1 M CaCl_2 aqueous solution at room temperature for 2 days followed by filtering and repeated washing in deionized water. The particles were then suspended in water, centrifuged and lyophilized for 2 days. Aldehydes may also be used as cross-linking agents.

5 (3) Uniform crosslinked gelatin microparticle fabrication.

A gelatin solution was obtained by dissolving 1 g of gelatin in 20 mL of deionized water at 40 °C . The solution was supplied to the main stream tank of the system described in Example 1. Canola oil was used for the carrier stream and the collection bath. The carrier stream solution was maintained at room
10 temperature, whereas the collection bath was kept at 0 to 4 °C.

A stable jet with a desired main stream diameter was obtained by applying the appropriate flow rates of the main stream and carrier stream, and the jet was broken into droplets by acoustic excitation.

While the droplets were being generated, a charging voltage of 0.5 to 1 kV
15 was applied, yielding charged droplets that spread in the collection bath. The temperature of the collection bath was maintained at 0 to 4 °C, and the droplets were gelled before aggregation.

The collection bath containing the gelatin gel particles was then stirred slowly for 1 hr while maintaining the low temperature. The same amount of cold acetone
20 as that of the oil was added to the collection bath and the resulting mixture was stirred for 30 min to extract water in gelatin hydrogel drops. The uniform gelatin microspheres were then filtered, washed with cold acetone several times and lyophilized for 2 days. Figure 5 illustrates the uniform size distribution of gelatin microparticles with an isoelectric pH of 5.0. Figure 6 illustrates the uniform size
25 distribution of gelatin microparticles with an isoelectric pH of 9.0.

20 mL of 0.1% solution of the surfactant TWEEN 80TM (Uniqema, New Castle, DE) was prepared and 500 μL of glutaraldehyde was added to the solution. The pH of the solution was adjusted depending on the isoelectric pH of the gelatin. The solution was then stored at 0 to 4 °C. After lyophilization, 20 mg of uniform

gelatin microspheres were suspended in the prepared glutaraldehyde solution and cured for 24 hr at 0 to 4 °C with stirring.

After crosslinking, the microspheres were filtered and washed thoroughly with deionized water. The microspheres were then suspended in 100 mM glycine solution and stirred at room temperature for 1 hr. The microspheres were filtered once more, washed thoroughly with deionized water and lyophilized. Figure 7 illustrates light micrographs of crosslinked microsphere. The particles of panels A and B had an original, pre-crosslinking diameter of 50 μm . The particles of panel C had an original diameter of 100 μm , whereas the particles of panel D had an original diameter of 150 μm .

(4) Droplet chopping with blades

A solution of ethyl cellulose and red rhodamine dye dissolved in methylene chloride was broken into 150- μm -diameter droplets via the acoustic and carrier stream method described above. The droplets were then chopped into droplets of two different sizes with a blade, where the two different sizes depended on the position of the blade tip, as shown by the black arrow. Figure (8A) illustrates the chopping of droplets into smaller droplets of respectively 142- μm -diameter and 80- μm -diameter. Figure (8B) illustrates the chopping of the droplets into smaller droplets of respectively 149- μm -diameter and 40- μm -diameter.